IN THE SPECIFICATION:

After the title please insert the following sub-title and paragraph:

CROSS REFERENCE TO RELATED APPLICATIONS

This application claims the benefit of PCT Application No. PCT/CH2004/000387, filed on June 24, 2004 and European Patent Application No. 03015012.2, filed July 2, 2003, the disclosures of which are herein incorporated by reference in their entirety.

Please insert the following heading prior to paragraph **[0001]**:

FIELD OF THE INVENTION

Please insert the following heading prior to paragraph **[0002]**:

BACKGROUND OF THE INVENTION

Please insert the following heading prior to paragraph **[0006]**:

SUMMARY OF THE INVENTION

Please amend paragraph **[0007]** as follows:

[0007] This object is achieved by an MR imaging method ~~with the features of claim 1~~. wherein a series of successive magnetic resonance signals is obtained by steady-state free precession imaging. The successive magnetic resonance signals in the series are acquired by successively scanning respective set points in k-space in an undersampled fashion. The magnetic resonance signals in the series are acquired in conjunction with an eddy current reduction technique, and successive magnetic resonance images are reconstructed from the successive sets of magnetic resonance signals using a suitable reconstruction method. A practical use of this invention, as described in detail below, is in accelerating cardiac cine 3D imaging in order to obtain a sequence of cine images in a much shorter time or with a higher temporal resolution for a fixed acquisition duration. Another promising use of this invention is in accelerating free-breathing, untriggered (referred to as real-time hereafter) imaging to achieve higher spatial and/or temporal resolutions.

Please insert the following heading prior to paragraph **[0010]**:

BRIEF DESCRIPTION OF THE DRAWINGS

Please amend paragraph **[0010]** – **[0019]** as follows:

[0010] Further advantages of the invention are disclosed in the ~~dependent~~ claims and in the following description in which an exemplified embodiment of the invention is described with respect to the accompanying drawings. ~~It shows~~

[0011] Fig. 1 shows a regular sampling pattern of magnetic resonance signals,

[0012] Fig. 2 shows a time series of fully sampled images 1 and corresponding signal distribution of object intensities 3 along the dashed line 2 after inverse Fourier transform over time,

[0013] Fig. 3 shows a schematics of the *k-t* BLAST method,

[0014] Figs. 4a-4d show a cine 3D SSFP: influence of sampling order on image artifacts,

[0015] Figs. 5a-5d show a real-time 2D SSFP: influence of sampling order on image artifacts,

[0016] Fig. 6 shows the sampling pattern for cine 3D *k-t* BLAST with integrated training data,

[0017] Figs. 7a-7c show a cine 3D single breath-hold data set obtained with 5x *k-t* BLAST in a volunteer,

[0018] Figs. 8a-8c show a cine 3D single breath-hold data set obtained with 5x *k-t* BLAST in a cardiac patient, and

[0019] Fig. 9 shows a diagrammatically a magnetic resonance imaging system.

Please insert the following heading prior to paragraph **[0020]**:

DESCRIPTION OF PREFERRED EMBODIMENTS

Please amend paragraph **[0020]** as follows:

[0020] The MR method according to the present invention is based on an approach for accelerated data acquisition in dynamic imaging, called *k-t* BLAST

(Broad-use Linear Acquisition Speed-up Technique) and *k-t* SENSE (SENSitivity Encoding). The *k-t* BLAST and *k-t* SENSE methods are described in an abstract byJ. Tsao et. al. (Prior-information-enhanced Dynamic Imaging using Single or Multiple Coils with *k-t* BLAST and *k-t* SENSE. Proc. Int. Soc. Magn. Reson Med., Abstract 2369. 2002).

Please amend paragraph **[0029]** as follows:

[0029] Using 5-fold undersampling (5x *k-t* BLAST), volumetric data sets covering the heart with 20 slices at a spatial resolution of 2x2x5 mm^3 were recorded with 20 cardiac phases in a single breathhold of 20-22 sec. with the training data acquired in a separate 5 sec prescan. When the training data were acquired together with the undersampled data in a single scan (referred to as interleaved acquisition hereafter), the breathhold duration was prolonged slightly to 25-27 sec. The central, densely-sampled region in the k_y-k_z plane consisted of 49 profiles providing low-resolution images without aliasing. A schematic of the sampling pattern applied is given in Figure 6. The central portion [[15]] of the k_y-k_z space is sampled at the full FOV in each cardiac phase whereas the outer portion [[16]] is under-sampled by a factor of five with the pattern being shifted as a function of cardiac phase ~~(Figure 6b)~~. The gray dots 24 indicate the sampled points, the dashed dots 25 represent points sampled in the previous cardiac phase. Remaining acquisition parameters were as follows: FOV: 320x210 mm^2, TE/TR: 1.55/3.1 ms, flip angle: 45 deg., heart phase interval: 28-35 ms. Taking the acquisition of low-resolution training data into account, the net acceleration factor was 4.3.

Please amend paragraphs **[0041]** – **[0042]** as follows:

[0041] The magnetic resonance imaging system includes a set of main coils 10 whereby a steady, uniform magnetic field is generated. The main coils are constructed, for example in such a manner that they enclose a tunnel-shaped examination space. The patient to be examined is slid on a table 34 into this tunnel-shaped examination space. The magnetic resonance imaging system also includes a number of gradient coils [[11, 12]] 31, 32 whereby magnetic fields exhibiting spatial

variations, notably in the form of temporary gradients in individual directions, are generated so as to be superposed on the uniform magnetic field. The gradient coils [[11, 12]] 31, 32 are connected to a controllable power supply unit 40, [[2]]41. The gradient coils [[11, 12]] 31, 32 are energized by application of an electric current by means of the power supply unit [[21]]. The strength, direction and duration of the gradients are controlled by control 40 of the power supply unit. The magnetic resonance imaging system also includes transmission and receiving coils [[13, 15]] 33, 36 for generating RF excitation pulses and for picking up the magnetic resonance signals, respectively. The transmission coil [[13]] 33 is preferably constructed as a body coil whereby (a part of) the object to be examined can be enclosed. The body coil is usually arranged in the magnetic resonance imaging system in such a manner that the patient [[30]] 50 to be examined, being arranged in the magnetic resonance imaging system, is enclosed by the body coil [[13]] 33. The body coil [[13]] 33 acts as a transmission aerial for the transmission of the RF excitation pulses and RF refocusing pulses. Preferably, the body coil [[13]] 33 involves a spatially uniform intensity distribution of the transmitted RF pulses. The receiving coils [[15]] 36 are preferably surface coils [[15]] 36 which are arranged on or near the body of the patient [[30]] 50 to be examined. Such surface coils [[15]] 36 have a high sensitivity for the reception of magnetic resonance signals which is also spatially inhomogeneous. This means that individual surface coils [[15]] 36 are mainly sensitive for magnetic resonance signals originating from separate directions, i.e. from separate parts in space of the body of the patient to be examined. The coil sensitivity profile represents the spatial sensitivity of the set of surface coils. The ~~transmission~~ receiving coils, notably surface coils, are connected to a demodulator [[24]] 44 and the received magnetic resonance signals (MS) are demodulated by means of the demodulator [[24]]. The demodulated magnetic resonance signals (DMS) are applied to a reconstruction unit 45. The reconstruction unit reconstructs the magnetic resonance image from the demodulated magnetic resonance signals (DMS) and optionally on the basis of the coil sensitivity profile of the set of surface coils. The coil sensitivity profile has been measured in advance and is stored, for example electronically, in a memory unit which is included in the reconstruction unit. The reconstruction unit 45 derives one or more image signals from the demodulated magnetic resonance signals (DMS), which image signals represent one or more, possibly successive magnetic resonance images. This means that the signal levels of the image signal of such a magnetic resonance image represent the brightness values

of the relevant magnetic resonance image. The reconstruction unit [[25]] 45 in practice is preferably constructed as a digital image processing unit [[25]] 45 which is programmed so as to reconstruct the magnetic resonance image from the demodulated magnetic resonance signals and optionally on the basis of the coil sensitivity profile. The digital image processing unit [[25]] 45 is notably programmed so as to execute the reconstruction in conformity with the present invention. The image signal from the reconstruction unit is applied to a monitor [[26]] 46 so that the monitor can display the image information of the magnetic resonance image (images). It is also possible to store the image signal in a buffer unit [[27]] 47 while awaiting further processing, for example printing in the form of a hard copy.

[0042] In order to form a magnetic resonance image or a series of successive magnetic resonance images of the patient to be examined, the body of the patient is exposed to the magnetic field prevailing in the examination space. The steady, uniform magnetic field, i.e. the main field, orients a small excess number of the spins in the body of the patient to be examined in the direction of the main field. This generates a (small) net macroscopic magnetization in the body. These spins are, for example nuclear spins such as of the hydrogen nuclei (protons), but electron spins may also be concerned. The magnetization is locally influenced by application of the gradient fields. For example, the gradient coils [[12]] apply a selection gradient in order to select a more or less thin slice of the body. Subsequently, the transmission coils apply the RF excitation pulse to the examination space in which the part to be imaged of the patient to be examined is situated. The RF excitation pulse excites the spins in the selected slice, i.e. the net magnetization then performs a precessional motion about the direction of the main field. During this operation those spins are excited which have a Larmor frequency within the frequency band of the RF excitation pulse in the main field. However, it is also very well possible to excite the spins in a part of the body which is much larger ~~man~~ than such a thin slice; for example, the spins can be excited in a three-dimensional part which extends substantially in three directions in the body. After the RF excitation, the spins slowly return to their initial state and the macroscopic magnetization returns to its (thermal) state of equilibrium. The relaxing spins then emit magnetic resonance signals. Because of the application of a read-out gradient and a phase encoding gradient, the magnetic resonance signals have a plurality of frequency components which encode the spatial positions in, for example the selected slice. The *k*-space is scanned by the

magnetic resonance signals by application of the read-out gradients and the phase encoding gradients. According to the invention, the application of notably the phase encoding gradients results in the sub-sampling of the *k*-space, relative to a predetermined spatial resolution of the magnetic resonance image. For example, a number of lines which is too small for the predetermined resolution of the magnetic resonance image, for example only half the number of lines, is scanned in the *k*-space.